Transthoracic Ventricular Defibrillation in the Dog with Truncated and Untruncated Exponential Stimuli

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Abstract—From 10 560 transthoracic fibrillation–defibrillation episodes in large anesthetized dogs, the effectiveness of 88 types of untruncated and truncated exponential waveforms in reversing ventricular fibrillation was evaluated. The study involved waveforms which could be generated with stored energy levels (in the simple capacitor–switch sense) of 60, 90, 120, and 180 J and initial current levels of 10, 20, 30, 40, 60, 80, and 100 A. The 10-A waveforms were untruncated or truncated at final current values of 5, 7.5, and 9 A. The 20-, 30-, and 40-A waveforms were untruncated or truncated at 5, 10, and 15 A. The 60-, 80-, and 100-A waveforms were untruncated or truncated at the 15-A level.

Optimally truncated waveforms with initial currents of 10, 20, and 30 A and for all stored energy levels were very much superior to the corresponding untruncated waveforms. Optimally truncated waveforms with initial currents of 20, 30, and 40 A and a stored energy level of 180 J were 98 to 100 percent successful in reversing fibrillation. Waveforms with an initial current of 10 A were much less effective, and those with initial currents of 60, 80, and 100 A were moderately less effective than our optimal waveforms.

Introduction

After Kouwenhoven *et al.* reported in 1960 that effective cardiac massage could be carried out on an external basis by the application of periodic pressure to the sternum [1], an alternating current transthoracic defibrillation system which Kouwenhoven *et al.* had developed earlier [2] came into widespread clinical use. In this system, a 480-V 60-Hz sinusoidal 0.25-s shock is furnished by a step-up transformer and mechanical relay arrangement. Power is furnished by the line as it is delivered to the patient's chest. In a 1962 paper, Lown *et al.* reported that a waveform which could be generated by discharging a capacitor through a series inductor and the patient's chest was very effective in achieving ventricular defibrillation [3]. In this system, the storage capacitor is charged at a relatively slow rate from the ac line by means of a step-up transformer and rectifier arrangement or from a battery and a dc-to-dc converter arrangement. The stored energy is then delivered in a short pulse to the patient's chest when a mechanical relay is activated. At the present time, this is the most widely used clinical defibrillation system. Another widely used system, as suggested by Balagot *et al.*, involves storing energy in the several capacitors of a lumped delay line which consists of both capacitors and inductors and the use of a mechanical relay for discharging the delay line into the patient's chest [4].

Defibrillation devices which depend upon an inductor or inductors for pulse shaping and a mechanical relay for switching tend to be large and heavy. Typically, such devices have been designed to be moved to the patient's bedside on a cart, utilized for fixed operation in a coronary care unit, or used in an ambulance.

Recently, there have been attempts to develop lighter weight and more portable defibrillators. For example, in one such device[1] which weights $23\frac{1}{2}$ lbs, pulse shaping is achieved with solid-state switches which furnish a trapezoidal waveform by both initiating and terminating the discharge of an energy storage capacitor. Our group had previously demonstrated and reported that appropriate trapezoidal waveforms were very effective in achieving transthoracic ventricular defibrillation [5].

However, the trapezoidal waveform approach fails to fully exploit the energy storage capabilities of the capacitor in that only a small fraction of the energy originally stored in the capacitor is delivered to the patient. One way of more fully utilizing the storage capability of a capacitor, and consequently achieving a smaller defibrillator, is to let the capacitor potential decay further and thus generate either a truncated or an untruncated exponential waveform.

The present paper is concerned with a systematic experimental study of the effectiveness of truncated and untruncated exponential stimuli in achieving transthoracic ventricular defibrillation in large dogs. In addition, consideration is given to how our extensive data for transthoracic defibrillation might be extrapolated to predict probable results with electrode systems which are currently being considered for use in conjunction with very small totally implanted standby defibrillation systems.

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[1] Medical Research Laboratories, Inc., Model 500-B/LP.

Methods

Selection of Waveforms to be Studied

A representative waveform of current is shown in Fig. 1. Any given waveform is completely and uniquely specified by means of the initial current I_0, the time constant of decay τ, and the final current I_f.

Experimental data were collected to allow us to plot families of curves of percent successful defibrillations as a function of final current for initial current levels of 10, 20, 30, 40, 60, 80, and 100 A. Four curves, representing as many time constants of decay, were included in each family.

For each family of curves, the time constants were selected so that the energy levels associated with the untruncated waveform point on each curve ($I_f=0$) were 60, 90, 120, and 180 J. These energy levels also represent the energy which would have to be stored in a capacitor if one visualizes a waveform as being generated by simply closing and opening a switch to initiate and terminate the capacitor discharge through the chest.

The energy content in joules of an untruncated exponential waveform is given by

$$U_0 = 0.5\tau I_0^2 R \tag{1}$$

where τ is expressed in seconds, I_0 in amperes, and R (the chest resistance) in ohms. For the large animals and for the electrode configuration used in our experimental work, 60 Ω is a representative value for chest resistance. On the basis of this value of chest resistance and (1), the time constant of decay is related to the initial current and the energy by

$$\tau = \frac{U_0}{30 I_0^2} \tag{2}$$

and this expression may be used to calculate the time constants which were used in our study.

Procedure

A colony of 6 mongrel dogs, weighing from 24 to 32 kg, was maintained for our experimental work. Normally, an animal was removed from the colony and replaced by another one only after it had been subjected to 300 fibrillation–defibrillation episodes. An occasional animal was replaced earlier because of apparent central nervous system damage or death or was carried beyond the 300-episode mark because of the temporary unavailability of a replacement. A total of 58 animals was used in our study.

In the evaluation of a given waveform of current, an animal was chosen from the colony and anesthetized with pentobarbital sodium at 27.5 mg/kg of body weight. Additional anesthesia was often required during the procedure. The animal was placed on a thick Plexiglas table top and two chest electrodes, 9 cm in diameter, were taped to the anterior portion of the chest. One of the electrodes was placed approximately over the apex of the heart and the other one was positioned

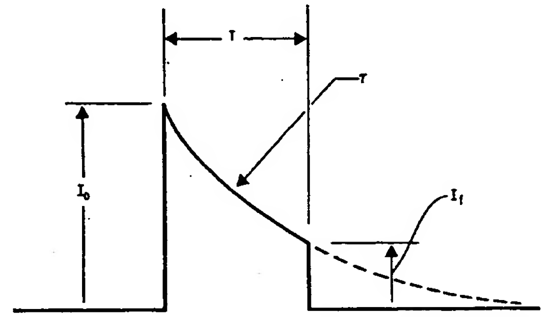

Fig. 1. Truncated waveforms. In the untruncated decay, the current is not abruptly terminated, but allowed to approach zero in an asymptotic fashion as suggested by dashed curve.

slightly to the right of midline and somewhat higher on the chest.

Ventricular fibrillation was then induced by a low-current shock. Thirty seconds later an attempt to defibrillate was made with the waveform being evaluated. If defibrillation was achieved on the initial trial, the episode was recorded as a success and the electrocardiogram was recorded for a 2½-min period. In the event of failure on the initial trial, a shock of known high effectiveness was used to salvage the animal and the episode was recorded as a failure. In either event, the procedure was repeated with not less than 3 min between the start of successive episodes. The animal was carried through a series of 20 episodes.

Five more animals were then taken through identical series. Thus, the percent effectiveness determination for each waveform was based on a total of 120 trials, 20 on each of 6 animals. The animals were ordinarily reused in the evaluation of additional waveforms with one or more days always intervening between successive procedures on a given animal.

All the shocks for inducing fibrillation and the defibrillatory shocks which had an initial current of 10 A and time constants of 40 and 60 ms were supplied by a special vacuum tube amplifier which has been described elsewhere [6]. Defibrillatory shocks which had an initial current of 10 A and time constants of 20 and 30 ms and all of the defibrillatory shocks with initial currents of 20, 30, 40, 60, 80, and 100 A were supplied by a hydrogen thyratron defibrillator [7]. The highly effective follow-up shock which was used to salvage the animal in case the shock being tested failed to defibrillate was furnished by the vacuum tube amplifier.

Results

In Figs. 2–8 the percent successful defibrillation is related to the final current for the 28 different initial current–time constant combinations which were studied. Since each of the 88 plotted points involves 120 trials, the figures summarize data on 10 560 fibrillation–defibrillation episodes.

In these figures, untruncated exponential waveforms have final current values which approach zero; trun-

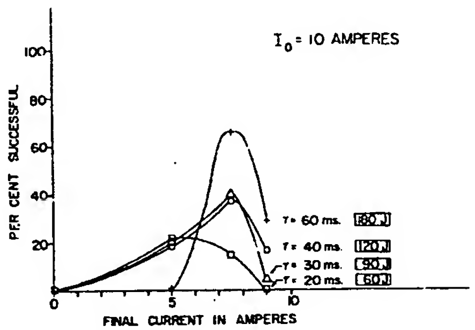

Fig. 2. Relation between percent success of ventricular defibrillation and final current of exponential stimuli with initial current of 10 A.

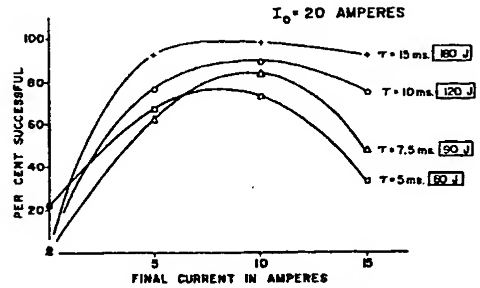

Fig. 3. Relation between percent success of ventricular defibrillation and final current of exponential stimuli with initial current of 20 A.

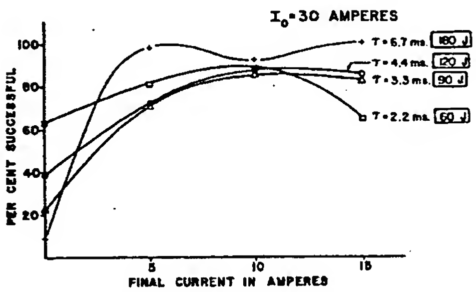

Fig. 4. Relation between percent success of ventricular defibrillation and final current of exponential stimuli with initial current of 30 A.

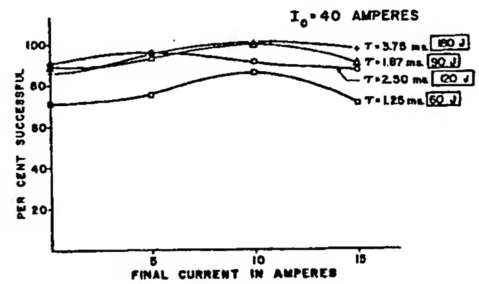

Fig. 5. Relation between percent success of ventricular defibrillation and final current of exponential stimuli with initial current of 40 A.

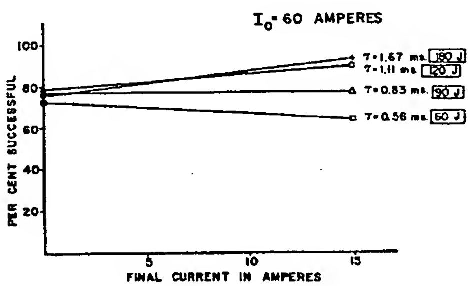

Fig. 6. Relation between percent success of ventricular defibrillation and final current of exponential stimuli with initial current of 60 A.

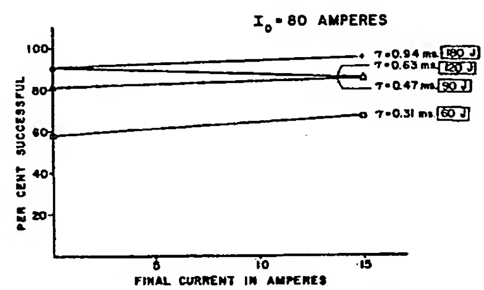

Fig. 7. Relation between percent success of ventricular defibrillation and final current of exponential stimuli with initial current of 80 A.

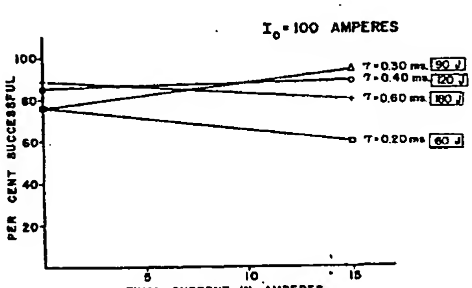

Fig. 8. Relation between percent success of ventricular defibrillation and final current of exponential stimuli with initial current of 100 A.

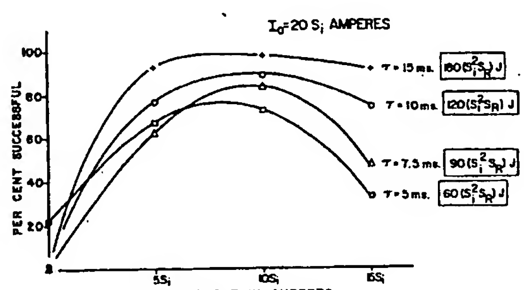

Fig. 9. Illustration of how Fig. 3 may be modified to apply to electrode system in which electrodes are surgically implanted on the rib cage.

cated waveforms have nonzero final current values. The actual duration of the shock in seconds associated with each of the points for the truncated waveforms is given by

$$T = \tau \ln (I_0/I_f) \tag{3}$$

where τ is also expressed in seconds. Since the current in the untruncated waveforms approaches zero in an asymptotic fashion, it is impossible to specify definite durations for these waveforms.

Discussion

The curves of Figs. 2–5 have a common feature in that the effectiveness of defibrillation increases, reaches a maximum value, and then decreases as the final current value increases.

From our study of triangular and trapezoidal waveforms [5] as well as from the data presented in the present paper, we believe that the reduced effectiveness associated with the untruncated waveforms (final current zero) is due to a refibrillation process which takes place as the current sweeps through very low values. That is, we believe that the leading portion of the exponential discharge defibrillates the heart while the "tail" of the discharge effectively refibrillates the heart. In comparing the families of curves with initial currents of 10 through 40 A, it is apparent that the effectiveness of the untruncated waveforms tends to increase with increasing values of initial current and consequent reduction in the time constant of decay. We attribute the very low effectiveness of untruncated waveforms with time constants above about 5 ms to the fact that the current sweeps through the critical refibrillation region for a comparatively long period, and the more acceptable effectiveness of untruncated waveforms with smaller time constants to the fact that the current sweeps through the critical region much more rapidly.

The decrease in effectiveness, which is observed in the families of curves with initial current values of 10 through 40 A when the final current exceeds some optimal value, is caused by a decrease in the energy which is actually delivered to the chest. The energy delivered to the chest by a truncated waveform U is related to the energy content of the corresponding untruncated waveform by

$$U = U_0\left[1 - \left(\frac{I_f}{I_0}\right)^2\right]. \tag{4}$$

The reduction in effectiveness at a final current of 15 A appears to be much more pronounced for the family of curves with an initial current of 20 A than for the family with an initial current of 40 A. This is because while $U=0.44U_0$ for the family of curves with an initial current of 20 A, $U=0.86U_0$ for the family of curves with an initial current of 40 A.

From the family of curves shown in Fig. 2, it is apparent that exponential waveforms having initial currents of 10 A are not very attractive for reversing ventricular fibrillation. Our explanation is, in part, based upon some earlier work in which the effectiveness of 5-A unidirectional rectangular wave shocks was evaluated as a function of duration of shock [8]. These data showed that the effectiveness of 5-A rectangular shocks increases from zero to about 70 percent as the duration of the shock varies from 2 to about 30 ms and then decreases rather rapidly with increasing duration of shock. We associate the relatively poor performance of the exponential shocks with time constants of 20, 30, and 40 ms and a final current of 5 A with the general ineffectiveness of current in the vicinity of 5 A in achieving ventricular defibrillation. We further associate the very poor performance observed for shocks with a time constant of 60 ms and a final current of 5 A with the poor results which were observed with 5-A rectangular wave shocks when the duration appreciably exceeded 30 ms.

In Figs. 6–8, data for untruncated waveforms and waveforms which are truncated at 15 A are presented for initial currents of 60, 80, and 100 A. Because of the low time constants associated with these waveforms, we do not believe that the refibrillation phenomenon is important. Furthermore, because of the high initial current values, truncation at 15 A does not result in a significant decrease in delivered energy as compared to that delivered by the corresponding untruncated waveform. As a consequence of these considerations, we would have expected the lines joining the untruncated and truncated versions of each exponential decay to be relatively flat. Although our experimental data generally support our expectations, several of the lines suggest more variation than we had anticipated.

Taken as a whole, the data presented in Figs. 6–8 suggest that while exponential waveforms with initial currents in the 60- through 100-A range are relatively effective, one apparently does not realize the near 100-percent effectiveness which occurs with optimally truncated waveforms of lower initial current amplitude. This finding seems to be compatible with that of an earlier study which indicated that 80- and 100-A rectangular waveforms were not as effective as lower current rectangular shocks [9].

Extrapolation of Results to Other Electrode Systems

Very recently there has been considerable interest in the possibility of developing very small standby and automatic defibrillation systems which could be totally implanted within a patient [10]–[14]. These systems have utilized two disk electrodes implanted on the rib cage [10], [11], one electrode within the right ventricle and the other under the skin of the anterior chest wall [12], and a bipolar catheter arrangement [13], [14]. In addition, there continues to be interest in open-chest defibrillation in which electrodes are applied directly to the myocardium.

How, one might ask, could the extensive transthoracic data presented in this paper be used to predict the probable results of using truncated and untruncated exponential waveforms with these other electrode systems? To try to answer this question, let us consider a

system in which disk electrodes are surgically implanted directly on the rib cage. For such an arrangement, some fraction of the electrode current f_{rib} will actually pass through the heart. The remainder will bypass the heart. In the transthoracic system with electrodes on the surface of the chest, some fraction of the electrode current f will pass through the heart.

If i_{rib} and i are the electrode currents for rib cage and transthoracic electrode systems, respectively, the condition that these electrode currents yield the same current through the heart is met by making

$$i_{rib} f_{rib} = i f. \tag{5}$$

From (5), the rib-cage electrode current required to yield the *same* current through the heart as i A of transthoracic electrode current is given by

$$i_{rib} = \frac{f}{f_{rib}} i = S_i i \tag{6}$$

where S_i is the f/f_{rib} ratio and defined as the current scaling factor.

Experimentally, S_i may be found by using a unidirectional rectangular waveform of some convenient pulsewidth and amplitude and finding how effective the waveform is in achieving defibrillation on a transthoracic basis. Using the same pulsewidth, one then finds the current amplitude required to achieve the *same* defibrillation effectiveness with rib-cage electrodes. The ratio of the rib-cage electrode current to the transthoracic electrode current is S_i.

If we define the resistance scaling factor S_R as the ratio of the rib-cage electrode system resistance to the transthoracic electrode system resistance, we have

$$R_{rib} = S_R R. \tag{7}$$

Experimentally, the resistance values and hence S_R may be found by simply measuring the amplitude of the voltage pulse which corresponds to a given rectangular wave pulse in each of the systems.

The energy content of an untruncated waveform applied via a rib-cage electrode system is given by

$$U_{0(rib)} = 0.5\tau I_{0(rib)}^2 R_{rib} \tag{8}$$

where $I_{0(rib)}$ denotes the initial electrode current in amperes. Using (6) to express $I_{0(rib)}$ in terms of I_0 and (7) to express R_{rib} in terms of R, (8) becomes

$$U_{0(rib)} = 0.5\tau I_0^2 R(S_i^2 S_R) \tag{9}$$

which, from (1), may be written as

$$U_{0(rib)} = U_0(S_i^2 S_R). \tag{10}$$

By using (6) and (10), the data presented in Figs. 2–8 may now be made applicable to a rib-cage electrode system by simply multiplying all of the indicated current values by S_i and all of the indicated energy values by $S_i^2 S_R$. The time constants remain unchanged. Fig. 9 illustrates this modification for Fig. 3.

In the discussion so far, we have tacitly assumed that if the total current through the heart is kept the same with a rib-cage electrode system as with a transthoracic system, the current distribution within the heart will also be the same. To the extent that this assumption is correct, the extrapolation from transthoracic electrode system data to rib-cage electrode system data would appear to be rigorous.

Since it is likely that for a given total current through the heart, the current distribution within the heart would be substantially the same in the rib-cage system as in the transthoracic system, we would anticipate that the extrapolation suggested above would yield nearly correct results for the rib-cage system. We have found this approach to be useful in helping us design rib-cage electrode systems for totally implanted defibrillators [10], [11].

In the case of electrode systems which involve an electrode within the heart or for open-chest defibrillation with electrodes applied to the myocardium, the current distribution within the heart would differ substantially from that for transthoracic defibrillation. Consequently, there is no *a priori* reason for believing that the extrapolation discussed for the rib-cage electrode system would yield rigorous results for these other systems. Nevertheless, one can still experimentally evaluate S_i and S_R as previously discussed, and in a formal fashion modify the transthoracic families of curves to represent the other electrode systems. Under these conditions, S_i is no longer interpreted as f/f_{rib} but rather as simply the ratio of current amplitudes required to achieve a given level of effectiveness of defibrillation with the two electrode systems. Such an approach gives considerable insight into the probable effectiveness of these systems under various conditions and we have found it to be useful in our work with a bipolar catheter electrode system.

In a sense, we would suggest that $S_i^2 S_R$ might be considered an inverse figure of merit for a defibrillation electrode system: the lower the $S_i^2 S_R$ product, the more favorable the system is likely to be for achieving effective low energy defibrillation.

As a further example of the carry-over from one electrode system to another, it is of interest that the general ineffectiveness of untruncated stimuli with time constants in excess of about 5 ms, as reflected in Figs. 2–4, is compatible with work recently reported by Geddes *et al.* in which it was shown that the energy required for open-chest defibrillation with myocardial electrodes increased some 15-fold as the time constant of decay increased from 5 to 10 ms [15].

Conclusions

We conclude that appropriately truncated exponential waveforms of current with initial values of 20 through 40 A are very effective in achieving transthoracic ventricular defibrillation in large dogs. Untruncated shocks within this initial current range are, in general, significantly less effective than corresponding shocks which are properly truncated. Exponential waveforms with initial currents of 10, 60, 80, and 100

A, whether truncated or not, are less effective than our optimally truncated waveforms in the 20- through 40-A range.

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